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The Impact of Differing Instability Devices on Postural Sway Parameters

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The Impact of Differing Instability Devices on Postural Sway Parameters

by

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ABSTRACT

Single-limb balance training is an integral part of preventing and rehabilitating lower extremity injuries. Practitioners use instability devices to provide a progressive overload to an individual during single-limb balance training sessions. Previous investigations have shown that when using instability devices, differences may or may not exist in postural sway parameters during use depending on the specific devices being assessed. Thus, this investigation sought to examine differences between a commonly used foam pad and a novel instability device (block) in measures of postural sway.

This experiment consisted of 22 healthy individuals with no history of lower extremity injury and neurological disorders. Participants performed three single-limb static balance conditions on a force platform sampling at 120 Hz. Each condition contained three 20-second trials separated by thirty seconds. The mean center of pressure (CoP) values of the three trials in each condition were then compared using a within-subjects repeated measures analysis of variance.

After evaluating the results, statistically significant differences were seen in sway area between conditions ($f(2,42) = 5.28$, $p = 0.009$), with the control (9.64 ± 4.53 cm) being significantly lower than both the foam pad (13.05 cm ± 4.25 cm) and block (12.33 ± 3.37 cm). Statistically significant differences were seen in CoP path length between conditions ($f(2,42) = 5.52$, $p = 0.007$), with the control (67.51 ± 9.49 cm) being significantly lower than both the foam pad (74.36 cm ± 9.76 cm) and block (76.38 ± 14.84 cm). Maximal medial-lateral CoP displacements were significantly different between conditions ($f(2,42) = 6.24$, $p = 0.004$). Lower displacements were seen in the control (1.39 ± 0.20 cm), which was statistically different from both the foam pad (1.59 ±

0.24 cm) and block (1.53 ± 0.25 cm). Maximal anterior-posterior CoP displacements were not significantly different between conditions ($f(2,42) = 1.50$, $p = 0.23$).

In conclusion, this investigation provides supporting evidence that different instability devices may provide similar changes in postural sway parameters in comparison to control conditions. The novel block instability device used in this investigation may be used in a similar fashion to the traditional foam pad in both prevention and rehabilitation settings based on no differences being found between the two devices.

Keywords:

DEDICATION

This thesis is a tribute to the patients and physical therapists who have played a significant role in the field of physical rehabilitation. I am grateful for their contribution towards advancing the knowledge and understanding of this important area of healthcare. The dedication shown by these individuals towards their work is inspiring, and it is an honor to acknowledge their efforts through this thesis. This thesis is a testament to investigating possible improvements in this field, and I hope it will serve as a source of inspiration for future research.

I would also like to dedicate this thesis to the School of Kinesiology and Nutrition at The University of Southern Mississippi. I am grateful for the faculty and staff who have supported me throughout my research journey. Their constant encouragement, feedback, and expertise have been invaluable to the success of my research project. It is because of their hard work and commitment that I have been able to learn and grow as a researcher, and I am truly thankful for everything they have done for me.

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LIST OF ABBREVIATIONS

AP	Anterior-Posterior
BoS	Base of Support
CAI	Chronic Ankle Instability
CNS	Central Nervous System
CoM	Center of Mass
CoP	Center of Pressure
EMG	Electromyography
LLA	Lateral Longitudinal Arch
ML	Medial-Lateral
MLA	Medial Longitudinal Arch
SMD	Space and Motion Discomfort
TA	Transverse Arch

CHAPTER I: INTRODUCTION

Balance is described as the contradicting actions between internal and external forces that create the natural sway of a human (Winter, 1995). Postural sway and the ability to control the magnitude of sway will determine the individual's competency in postural control. The body's center of mass (CoM) is located about the navel, and the base of support (BoS) is the point of contact with the ground (Horak, 1987). If an individual is in a bipedal stance, the base of support is greater than that of a unipedal stance. The surface on which the individual stands is also considered when identifying the base of support. If a subject is standing on a flat floor, there is a greater base of support than standing on a narrow or short surface.

An example could be a gymnastic balance beam, half foam roller, slack block, or other instability devices. Postural sway comprises medial-lateral (ML) and anterior-posterior (AP) sway directions. The AP sway is the path of the center of pressure (CoP) in a front-to-back direction. However, ML sway is the right-left CoP trajectory. Sway is the result of antagonistic anterior and posterior lower leg muscle contractions (Alexander, 1994; Pollock et al., 2000).

The visual, somatosensory, and vestibular systems are crucial in postural sway. The visual system evaluates motion in the visual field, whether from an external object moving or movement of the body (Redfern et al., 2001). The speed of external movement is the determinant of its influence on postural sway (Redfern et al., 2001). The somatosensory system identifies and adapts to surface changes. Skin receptor cells in the feet provide sensory information to the central nervous system (CNS) (Kaas, 2004). Muscular contractions are then initiated to maintain postural control. The vestibular system is responsible for recognizing the orientation and rotation of the head (Lee, 2022). Any change in head position will stimulate the semicircular canals and otolith organs to trigger a neuromuscular response to keep the head position in line with the CoM (Lee, 2022).

The use of systems creates a neuromuscular response, which is classified into three different strategies. The strategies used for postural control are the ankle, hip, and step. These strategies are the body's defense for preventing falls. The ankle strategy is the first attempt to control postural sway (Winter, 1995). It is described as using the dorsiflexion and plantar flexion muscles to decrease the momentum of the CoP trajectory. The hip strategy will be implemented if the external force is greater than the ankle strategy can control (Winter, 1995). The hip strategy recruits the trunk and thigh muscles to position the CoM over the CoP when the force exerted on the body exceeds the allowable area of the BoS (Horak & Nashner, 1986; Winter, 1995). The step strategy is the last line of defense when preventing a fall. When the magnitude of external force exceeds the base of support, individuals must move in the same direction as the force to widen the base of support and restore postural control (Horak & Nashner, 1986).

The mechanism of the foot is a critical component of postural sway and the ability to maintain balance. The structure of the medial longitudinal arch (MLA) and its effects on balance have been studied for several years (Birinci & Demirbas, 2017; Fukano & Fukubayashi, 2009; Oatis, 2009). Several studies support that the magnitude of CoP trajectory varies amongst subjects with flat feet and high arches compared to subjects with average arch heights (Birinci & Demirbas, 2017; Fukano & Fukubayashi, 2009; Oatis, 2009). The difference in MLA heights changes the anatomy of the tibia and talocrural joint angle (Nawoczenski et al., 1995; D. S. Williams et al., 2001). We can assume pressure in the foot is distributed differently when there is a low or a high MLA compared to an average MLA height. Studies have found that a low or absent MLA maintains pressure equally across the bottom of the foot (Han et al., 2011). However, individuals with a high MLA have higher pressure values in the lateral forefoot (Simeonov, 2001). We can assume that the difference in pressure locations will produce a difference in the CoP displacement when evaluating postural sway. Previous and reoccurring injuries will increase an individual's postural sway.

An example of this is chronic ankle instability (CAI). When the ligaments surrounding the ankle are stretched, it is difficult for the joint receptors to identify the position of the joint (Garrick, 1977; Hertel, 2000). Plantar fasciitis is another example that affects postural sway. Suppose the plantar fascia is stretched too tightly due to a high arch. In that case, the Achilles tendon will be under constant strain, which hinders the efficiency of the gastrocnemius and soleus muscles (Ağırman, 2018; Cheung et al., 2006). The constant pull on the plantar flexion muscles explains the increased forefoot pressure in individuals with high MLA (Ağırman, 2018; Cheung et al., 2006; Gonçalves et al., 2017).

Since the CoP data points increase when the BoS is decreased, we can assume that a unilateral stance will produce more instability than a bilateral stance (Mancini & Horak, 2010). This is due to the CoM being located in the medial aspect of the trunk and equally supported by the two legs (Winter, 1995). When individuals stand on one leg, the CoM is still located in the medial part of the trunk, so we can assume that the CoP data points will shift to the ML direction (Winter, 1995).

The surface of the BoS determines the acceleration of sway velocity (Stanek et al., 2013). When an instability device is used, the somatosensory system has to recognize the difference in sensory information (Kaas, 2004). Standard instability devices are foam pads (Airex, Neurocom, half-foam roller, slack block), air-inflated devices (BOSU ball, DynaDiscs), and wobble boards. The instability device's density and elasticity determine the postural sway's magnitude (Boonsinsukh et al., 2020). For instance, if a foam pad had a high density, which means it is very firm, it would be easier to control balance than if the density of the foam pad was low and highly flexible (Boonsinsukh et al., 2020). The use of an instability device depends on the training intervention and the variables being tested. It has been supported that a balance training intervention improves postural control (Wortmann & Docherty, 2013). Several studies have observed the effects of training interventions on CoP displacement (Karlsson & Persson, 1997; Wortmann & Docherty, 2013).

Since postural control is enhanced with balance training, we can assume it can be used in rehabilitation. Most injuries require a rehabilitation process to regain musculoskeletal strength and neuromuscular function. Based on the research presented in this literature, training interventions will decrease the displacement of CoP due to the activation of the surrounding muscles to prevent falling (Rossi et al., 2013; Wortmann & Docherty, 2013). This recruitment of muscular contraction will eventually result in a faster and more accurate contraction. Training on an instability device is especially beneficial for injuries around the joints used for postural control strategies: ankle, hip, and step (Rossi et al., 2013). If training interventions involving an instability device are known for rehabilitation, the limiting factor is the type of instability device used. The research discussed in this paper observes the differences in sway parameters within different instability devices (Boonsinsukh et al., 2020; Hong et al., 2015; Stanek et al., 2013).

Our experiment examines the sway parameters in individuals in a unilateral stance while standing on different instability devices: foam and block. The foam pad is a flat, square pad commonly used in rehabilitation clinics. The block is a new device with a rectangular foam pad with a wooden piece on top, and it is the width and length of the foot. Based on the supporting research, we can assume that a narrower base of support (block) would produce a greater amount of sway. However, we can make a contradicting assumption that the block will have a lower sway velocity compared to the foam due to the wooden piece on the top of the device.

Hypotheses

H_1: There will be no statistically significant differences in sway parameters between the foam and the block.

This statement is supported by an experiment by Boonsinsukh et al. examining the difference between two instability devices (Boonsinsukh et al., 2020). Their study concluded that both foam pads had similar results for sway area, CoM acceleration, and time to regain stability. This particular source is relevant to our study due to the utilization and comparison of sway parameters between two foam pads.

H_A: There will be statistically significant differences in sway parameters between the foam and block.

Increasing the body's CoM results in an increase in instability. Simeonov conducted research that observed the difference in sway area and velocity between two instability devices of different heights. One instability device was three meters high, and the other was nine meters high. The results of Simeonov's study support our hypothesis since the taller instability device produced a greater magnitude of sway (Simeonov, 2001).

H_2: I hypothesize that there will be a statistically significant increase in sway parameters for the instability devices compared to the floor.

The somatosensory system is the body's way of adapting to surface change. When someone stands on the ground, their BoS is stable. However, if someone stands on a foam pad, their postural control will decrease due to raising the CoM and inhibiting the somatosensory receptors in the feet. Previous studies support this hypothesis by observing sway parameters when standing on the ground versus a foam pad. Janura et al. conducted an experiment that traced the CoP displacements between the floor and a foam pad (Janura et al., 2017). Results from this study coincide with our hypothesis.

H_A: There will be no differences in postural sway parameters between the ground and the instability devices.

This hypothesis is possible due to the dimensions needed for an instability device to increase postural sway. Stanek et al. found that air-formulated rubber instability devices produce more sway than foam instability devices (Stanek et al., 2013). The density of the foam pad plays a crucial role in determining the amount of postural instability (Hong et al., 2015). It is possible that the foam pads used in our project do not have the required dimensions to produce a significant difference in sway parameters when compared to the ground.

CHAPTER II: LITERATURE REVIEW

Balance

Postural Sway

Postural control maintains the body's vertical projection of the CoM over the BoS without major postural adjustments. Horak defines postural control as aligning the body's CoM to the base of support when gravity is present. Postural sway is the deviation from the mean (CoP) of the foot for a given trial (Guskiewicz & Perrin, 1996). We use postural sway to evaluate an individual's probability of maintaining balance in an upright position (Hertel, 2000). A key point made by Mancini and Horak is that several factors in the balance control and physiological systems impact postural control (Mancini & Horak, 2010).

Postural sway is measured by force plates that obtain the CoP data points, determining the average velocity (Alexander, 1994; Hertel, 2000; Mancini & Horak, 2010). This testing is called posturography and can be tested by a static or dynamic stance. Static posturography focuses on changing CoP data points while the subject stands in place (Mancini & Horak, 2010). Dynamic posturography measures the body's ability to maintain balance when external factors are manipulated (Mancini & Horak, 2010). An example would be making subjects stand on a moveable platform. When the surface shifts, the body will quickly adjust to maintain balance.

The change in CoP data points reflects the musculoskeletal, peripheral, and central nervous systems' ability to work in unison (Hertel, 2000; Mancini & Horak, 2010). The body maintains postural control by interpreting sensory feedback via afferent nerves. The visual, auditory, vestibular, and higher motor cortex contribute to the extent of an individual's postural sway (Horak, 1987, 1997; Mancini & Horak, 2010). The body uses sensory information to maintain the CoM over the BoS by making continuous, minor adjustments (Alexander, 1994; Pollock et al., 2000). The head, eyes, trunk, or limbs make these adjustments (Horak, 1987, 1996, 1997; Mancini & Horak, 2010). Research has also supported the idea that postural sway is affected by the anticipation of movement (Horak, 1987). There is an increase in CoP data points when an individual anticipates a change in stability. This action is described as a feedforward mechanism(Horak, 1987).

In an attempt to make static posturography testing more difficult, sensory feedback can be altered. By teasing out the effectiveness of proprioception, the BoS, visual feedback, or auditory information, the balance system will not be able to work efficiently (Mancini & Horak, 2010).

Anterior-Posterior Sway.

Sway in the anterior-posterior (AP) direction is the CoP trajectory path length in the forward-backwards direction. AP ankle sway is controlled by the tibialis anterior, gastrocnemius, and soleus muscles. When the CoM exceeds the BoS in the anterior or posterior direction, the plantarflexors or dorsiflexion muscles will activate to regain postural equilibrium. When comparing the AP ankle sway data for the firm surface, feet-apart testing condition to the firm surface, feet-together condition in a study by Promsri, Haid, & Federolf (2020), there was a significantly higher result in AP ankle sway data points for the feet-apart condition ($p < 0.001$; $d = 0.79$). However, when placed on a modified balance board in a narrow stance, the AP sway was significantly lower than the AP sway results in the other conditions. This suggests that this testing condition was easier to maintain postural equilibrium since the displacement acceleration was lower

(Promsri et al., 2020). As expected, a wider stance will result in AP instability rather than ML. This could be due to the plantar flexion and dorsiflexion muscles having a lower activation rate due to reduced ankle mobility (Schmidle et al., 2022)

Medial-Lateral Sway.

In the ML direction, the CoP trajectory path length is in the medial-lateral (right to left) direction. When the body makes an ML shift in displacement, the peroneus muscle group around the ankle is activated. A study by Promsri, Haid, & Federolf (2020) observed the difference in CoP data when the postural control system is challenged by changing the BoS and bipedal stance width. When the subjects stood on a firm surface with their feet together, the ML ankle sway was significantly higher than when the subjects stood in the broader stance ($p < 0.001$; $d = 1.30$). As one would expect, a narrower stance will produce more instability in ML direction as the BoS is reduced in that direction.

Systems

Visual System

According to M.S. Redfern et al., the visual system plays a significant role in maintaining balance. The retina receives information from objects moving in the visual field or the body moving in space, defined as self-motion (Redfern et al., 2001). When objects in the visual field move, the body can experience postural changes, disequilibrium, and motion sickness. An example of how the body uses the visual system to maintain balance could be a subject standing statically in a room with little to no movement in the visual field (Redfern et al., 2001). It is discussed by M.S. Redfern et al. that the frequency range of movement needed to increase postural sway is less than 0.1

Hz. This research team used a slow-moving bus as an example. When a bus moves slowly in the subject's visual field, it will increase postural sway in the subject. In contrast, several studies support the idea that the movement speed in the visual field does not affect postural sway velocity (Clément et al., n.d.; Lestienne et al., 1977; Peterka & Benolken, 1995). There appears to be a lack of consistent findings in the literature to support the body moving in the same or opposite direction of moving object(s) in the visual field (Clément et al., n.d.; Lestienne et al., 1977; Peterka & Benolken, 1995). The threshold and saturation phenomenon are discussed in research by Peterka and Benolken in which the visual system has a saturation level. If the speed of the moving object is within a range, the body's sway velocity will not be affected. However, if the object's speed is outside of the range, the body's sway will adjust at a faster or slower velocity (Peterka & Benolken, 1995). This research team collected CoM data for normal and vestibular-impaired patients at different visual movement frequencies (0.1 Hz, 0.2 Hz, 0.5 Hz). One testing condition was a fixed, stable ground; the other was a rotating platform (sway-referencing). Results showed that the sway-referenced condition produced the most CoM displacement in the control group as well as the vestibular loss group (-87 ± 17 cm and -81 ± 15 cm, respectively) when the frequency of the visual surround motion was at 0.5 Hz compared to the fixed condition of -37 ± 32 cm (control subjects), and -16 ± 17 cm (vestibular loss subjects) (Peterka & Benolken, 1995). The data from this study may be evidence that supports the enhancement of functioning systems when one is impaired. Bednarczuk et al. evaluated the visual system's role when standing statically to support this claim further (Bednarczuk et al., 2021). After completing their study, subjects with no visual impairment showed a significant difference in balance between the eyes open and eyes closed conditions and when in a bipedal or unipedal stance. However, when subjects had visual impairments, there were no significant differences between the eyes opened and closed conditions in the bipedal stance. The loss of the visual system requires an enhancement of the somatosensory and vestibular systems to control postural sway (Bednarczuk et al., 2021). Research by M.S. Redfern et al. also supports this idea. An increased movement in the visual field will increase postural sway, whereas a fixed visual environment will decrease postural sway (Redfern et al., 2001).

If a subject suffers from a disorder, they may rely on the visual system more than the vestibular or somatosensory systems, depending on the medical condition. Another study by Redfern and Furman supports that visible movement increases postural sway in individuals with vestibular impairment (Redfern & Furman, 1994). This experiment observed postural sway differences when looking at a visual stimulus at different frequencies. The subject pool consisted of patients who suffered from vestibular impairment and individuals with no symptoms of vestibular impairment. Results defend the idea that vestibular impaired individuals have increased postural sway parameters. The body will overcompensate for this reduced function by increasing the use of other systems. This can cause visual sensitivity in patients with vestibular impairments. Visual sensitivity creates overstimulating environments for some individuals (Redfern et al., 2001; Redfern & Furman, 1994). Another term that describes visual acuity is known as “space and motion discomfort (SMD)” (Jacob et al., 1993; Redfern et al., 2001). SMD correlates to balance and panic disorders. Since individuals are sensitive to visual feedback, some environments, like cliffs or heights, could be frightening (Jacob et al., 1995; Redfern et al., 2001).

Redfern et al. (2001) explain that the visual system, in unison with the proprioceptive and vestibular systems, is a crucial factor in one's ability to maintain balance (Redfern et al., 2001).Their research also explains the importance of the visual system when other methods, like the vestibular system, are impaired. Knowing that the visual field primarily contributes to postural adjustments is imperative when researching postural sway. Researchers will instruct subjects to maintain visual contact with a specific mark or object. This has the effect of helping regulate visual distraction. Testing is conducted in controlled environments to decrease the amount of movement in the visual field.

Vestibular System

The vestibular system consists of the semicircular canals and the otolith organs in the ears. Three semicircular canals monitor head rotation. Each canal forms a 90-degree angle with the other two. This heightens sensitivity within the canals and allows the vestibular system to dictate which direction the head is rotating (Lee, 2022). The two otolith organs, the utricle and saccule, monitor increases in movement within a straight line (Lee, 2022). The saccule recognizes motion in the sagittal plane, as in up-down movements (Lee, 2022). The utricle detects movement in the horizontal plane. Examples of this movement could be left-right, forward-backward, or both (Lee, 2022).

The vestibular system is categorized as motor and sensory systems (Cherng et al., 2021; King & Horak, 2014). When considering the vestibular as a sensory system, the vestibular system works with the somatosensory and visual systems to maintain postural control. The central nervous system can then evaluate the environment (King & Horak, 2014). The vestibular organs control the head and trunk orientation through the vestibulospinal tract as a motor system (Cherng et al., 2021; King & Horak, 2014). It is discussed by King & Horak that the vestibular system is an essential concept for postural control in four aspects. The first role of the vestibular system is the recognition of the body's bearings and self-motion. Another role is maintaining the trunk in a vertical orientation. Sustaining the body's CoM over the BoS during static and dynamic movements is another crucial role of the vestibular system. It also decreases head instability during postural adjustments (King & Horak, 2014).

Like the other systems, the vestibular system can be manipulated to test postural sway. Horak and Hlavacka studied the effects of vestibulospinal sensitivity based on somatosensory loss (Horak & Hlavacka, 2001). This study used galvanic vestibular stimulation to manipulate the vestibular system. It is discussed that the vestibular system dictated whether the surface was stable or unstable and sent sensory information to make postural adjustments (Horak & Hlavacka, 2001; Mergner & Rosemeier, 1998).

Another way to hinder the vestibular system's function would be to spin the subject before testing postural sway parameters (Faquin et al., 2018). Faquin et al. provided evidence that vestibular manipulation causes an increase in task error and spatial disorientation. The literature explains that head rotation will lead to the body walking in that direction. The individual will take more steps to correct this reaction (Faquin et al., 2018).

Manipulation of the vestibular system has supported the pertinence of the vestibular system in balance and postural control. By making these alterations, results have shown that the efficiency of the given task is decreased.

Somatosensory System

The somatosensory system is our body's way of identifying objects and surfaces (Kaas, 2004). Afferent neuron receptors are aroused after skin cell receptors contact an object. Activation of the Golgi tendon organs and muscle spindles can also cause arousal of afferent receptors (Kaas, 2004). The afferents relay sensory information to the spinal cord and brain stem. Stimulating the somatosensory neurons in the lower brain stem and thalamus then activates the somatosensory area of the anterior parietal cortex (Kaas, 2004). Finally, the signal reaches the posterior parietal cortex, which houses the functions of attention, somatosensory motor functions, identification, memory, and motor fields of the frontal lobe (Kaas, 2004). This literature stated that this pathway controls muscular contractions through connections to the basal ganglia, sensory nuclei of the thalamus, and motor components of the brain stem and spinal cord (Kaas, 2004).

The somatosensory system is the most significant in maintaining postural control on unstable surfaces (Tanaka & Uetake, 2005; Wu & Chiang, 1997). To test the efficiency of the somatosensory system, many researchers, Kerr et al., Lord et al., Vuillerme et al. (2001), and Vuillerme et al. (2004), used a foam pad to study the differences in postural control before and after manipulation (Lord et al., 1991; Vuillerme et al., 2001; Vuillerme & Nougier, 2004). Standing on an instability device, such as a foam pad, alters the function of the joint receptors and the cutaneous mechanoreceptors in the foot (Wu & Chiang, 1997). Tanaka & Uetake altered the somatosensory system by testing the difference in postural sway parameters when subjects stood on a firm surface or a foam pad(Tanaka & Uetake, 2005). They used two different testing conditions: eyes-open and eyes-closed. They instructed the subjects to stare at a square on the wall before them to eliminate visual system manipulation. This study displayed an increased sway velocity in the AP direction on the foam pad regardless of the visual condition (Tanaka & Uetake, 2005). The results also showed a significant p-value for the foam pad-eyes open testing condition. The visual and vestibular systems will overcompensate due to the manipulation of the somatosensory system. The literature mentioned supports the idea that as objects and surfaces change, the somatosensory system will attempt to compensate for the instability.

Strategies

Ankle Strategy

Humans can recover from postural imbalances by using the ankle strategy. This strategy is the body's first line of defense in maintaining the CoM over the BoS. The ankle strategy is used when the individual is standing on a flat surface (Winter, 1995). Runge et al. described this strategy as an inverted pendulum with the most weight and broadness above a narrow support base (Runge et al., 1999).

Torque at the talocrural joint produces sway in the AP direction when in a bilateral stance (Runge et al., 1999). The plantarflexion and dorsiflexion muscles work antagonistically to control the inverted pendulum (Winter, 1995). Research was conducted to study the activation of the plantar flexion (gastrocnemius and soleus) and dorsiflexion muscles (tibialis anterior) in reaction to a forward-backward moving platform (Winter, 1995). Since these muscles are also used for inversion and eversion, it is evident that CoP moves medially while shifting the CoM anteriorly (Winter, 1995).

Karlsson & Persson observed how much ankle strategy was used in a bilateral stance for different conditions (Karlsson & Persson, 1997). They asked the subjects to stand as static as possible for quiet standing. Subjects were then instructed to repeat the quiet standing condition but with their eyes closed. Another condition in this study was the ankle strategy. The researchers instructed the subjects to sway forwards-backwards while keeping their feet planted. They developed two methods to obtain the body's CoM: model and marker. The model method uses formulas to give an estimation of the CoM. Karlsson and Persson used binary images and marker data from centroids to determine the CoM's location, called the marker method. The results of this study show that sway was low for all subjects in the quiet standing condition (Karlsson & Persson, 1997). The ankle strategy condition showed a continuous decrease in correlation within the estimated and actual acceleration values of the CoM. Although this study used a small pool of subjects, the researchers supported the idea that the ankle strategy is used more when the tasks become more challenging (Winter, 1995).

Muscles involved in the ankle strategy are recruited distal-proximal (Horak & Nashner, 1986). Since the body resembles an inverted pendulum, once the anterior-posterior sway travels farther than the surface on which the individual stands, the body must recruit a higher demanding strategy to achieve postural control.

Hip Strategy

The hip strategy is the body's second adaptation mechanism when faced with postural instability. When an individual is unstable, the hip strategy will overcompensate if the ankle strategy cannot adjust accordingly. Horak and Nashner state that balance is maintained by the hip strategy when the BoS is shorter than the length of the foot (Horak & Nashner, 1986). An example of this could be when a gymnast performs a balance beam routine. If the athlete's CoM travels farther than the width of the beam, she will activate the needed trunk and thigh muscles to regain postural control.

Literature by Winter (1995) investigated postural control when standing and walking by observing a subject standing on a platform that could shift forwards or backward (Winter, 1995). Electromyograph technology measured contractions of the leg muscles during this test. The results showed that when the platform shifted backward, the hip flexors used for the hip strategy dominated the muscles used for the ankle strategy. The force of hip flexors or extensors is much more than that of the ankle plantar flexors and dorsiflexors, which would shift the CoM more efficiently (Winter, 1995). Hip flexion shifts the CoM posteriorly, and extension of the hip will result in the CoM shifting anteriorly (Winter, 1995). Unlike the ankle strategy, the hip strategy recruits muscles proximal-distal relative to the hip joint (Horak & Nashner, 1986).

The subject's stance is a crucial aspect of the role of the hip strategy. The study conducted by Winter concluded that the hip strategy controls sway in the ML direction during quiet bipedal standing (Winter, 1995). However, it stated that the hip strategy controls sway in the AP direction while in a tandem stance (*Figure 1*). When an individual is walking, the hip flexors and extensors are responsible for CoM in the AP direction, and the hip abductors control the medial-lateral sway of CoM.

Research supports that the hip strategy is effective when the surface is not slippery since the mechanism contacts the ground with a horizontal force (Horak & Nashner, 1986). Once the sway velocity or distance exceeds the hip strategy range, another mechanism must be recruited to prevent falling.

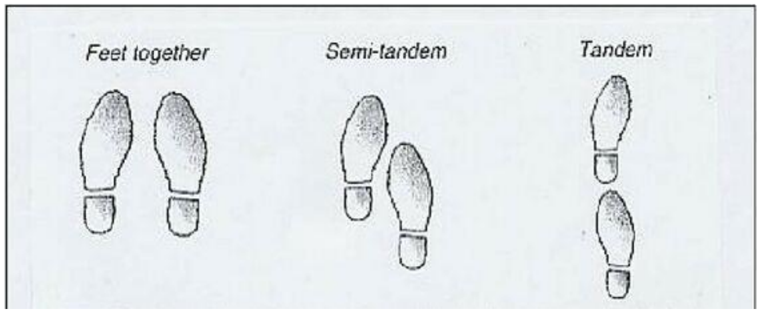

***Figure 1**. Illustration of Stances.*

Step Strategy

The step strategy is implemented when the force or acceleration of postural sway demands more adjustment than the hip strategy can exert (Horak & Nashner, 1986). An individual will take a step that repositions the base of support back underneath the CoM, restoring balance equilibrium. The step for recovery can be made in any direction, depending on the movement of the force.

For example, a study by Lee et al. observed the step strategy by using a platform that shifted forward and backward (Lee et al., 2014). This study aimed to discover the difference in step strategy concerning age. Within age groups, the forward and backward protective steps were compared to a voluntary command of the steps. For the protective step strategy, subjects stepped in the opposite direction of the moving platform. As a control, the researchers asked the subjects to step forward and backward when the stimulus was given. Results showed that backward protective steps were smaller in distance and had a shorter reaction time than forward protective steps. The forward trunk motion when the platform is shifted backward could factor in the longer reaction time for the forward protective step.

When balance is disturbed by a lateral force, the individual will step medially or laterally to reestablish equilibrium of the CoM. Mille et al. organized research investigating the lateral step strategy within different age groups (Mille et al., 2005). This study used a waist-pull system to disturb equilibrium. It is explained that there are three types of ML step strategies: loaded side step and unloaded crossover step (Mille et al., 2005). An example of the loaded side step strategy would be the waist-pull system pulling the subject to the left, and the subject takes a lateral step with the left leg. An unloaded crossover step would occur when a subject is pulled to the left and takes a crossover step with the right leg. Results displayed that older adults tend to use the unloaded crossover step, which requires additional steps to recover balance. This strategy also raises the risk of an individual tripping due to the crossover steps (Mille et al., 2005). The loaded side step strategy can be deemed more efficient because it results in a faster step duration with a shorter step needed.

When the CoM is shifted anteriorly or posteriorly, the body must step in the opposite direction to recover balance. However, when the CoM is externally manipulated to the left or right, the individual should step in the same direction in which he or she is being pushed or pulled. Studies mentioned consider age and disease in their research. As humans age or obtain a disease, the recruitment of motor neurons and muscular strength tends to decrease. This can provide slower step strategies, increasing injuries or deaths due to falling.

Mechanisms of the Foot

Arches

The foot consists of three arches- the medial longitudinal arch (MLA), the lateral longitudinal arch (LLA), and the transverse arch (TA) (*Figure 2*). MLA covers most of the foot and is considered one of the most impactful arches. The MLA covers the first metatarsal, medial cuneiform, navicular, calcaneus, and talus (Birinci & Demirbas, 2017; Fukano & Fukubayashi, 2009; Oatis, 2009). This arch is measured by classifying individuals with having a low, normal, or high arch. One of the functions of the arches of the foot is to absorb shock from the ground. Individuals with a high MLA are at risk for specific injuries, including heel pain (Mølgaard et al., 2010). This is caused by having more tension on the plantar fascia, which pulls on the calf muscles through the connection of the Achilles tendon (Ağırman, 2018; Cheung et al., 2006; Richer et al., 2022). This constant pull on the gastrocnemius and soleus muscles will also create an imbalance, resulting in a higher sway velocity than someone with a low or average MLA height. It is assumed that people with high arches would have a higher sway velocity since there is less medial contact with the ground. Studies conducted by Chang et al. and Cote et al. support the idea that the different MLA heights directly correlate to postural control (Chang et al., 2010; Cote et al., 2005). Studies by Williams et al. and Nawoczenski et al. result in findings that suggest individuals with low MLA have a higher ankle eversion to internal rotation of the tibia ratio when running (Nawoczenski et al., 1995; Williams et al., 2001; Williams et al., 2014). These findings also lead to increased stress on the hip and knee joints. The increased proportion of ankle eversion to internal rotation of the tibia for people with low arches can also result in lateral ankle sprains.

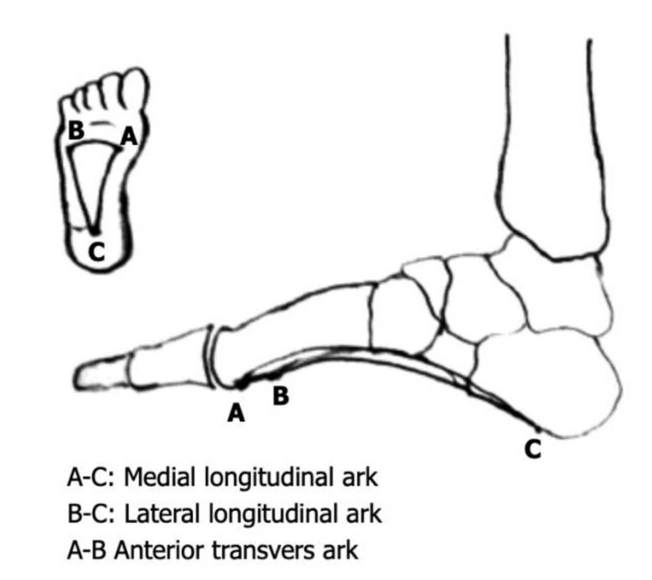

Figure 2. *Arches of the Foot* (Atik, 2014)

Research by Cote et al. investigated the difference in CoP excursion and postural sway parameters during a static one-leg stance among subjects with different foot structures. This study explains the difference in postural stability when the foot is excessively supinated (high arch) and pronated (low arch, flat-footed). Since MLA height can change the foot's structure, we can assume that the foot's mechanisms will also differ. Having a supinated or pronated foot demands an amplified usage of surrounding muscles due to the difference in the angle of the talocrural joint. If the foot has a low MLA or no MLA, the foot is pronated, causing the body weight from the tibia to exert a force on the medial side of the foot. If the MLA is high, there is less contact with the ground, and the force will be exerted on the lateral side of the foot. It can be assumed that a supinated foot has higher instability due to fewer plantar sensory receptors in contact with the ground (Cote et al., 2005). In this study, postural sway did not significantly differ across the normal, supinated, and pronated testing groups. However, there was a significant difference in the stability index of the pronator group compared to the supinator group. There was a high variability of stability index within the pronated testing group (0.0082 [variable]cm/[height]cm) and a lower variability of stability index in the supinated testing groups (0.0071 [variable]cm/[height]cm). This could have been due to the pronated foot having total contact with the ground, allowing the CoP to travel more. It also could be a result of the flexibility of the pronated foot. Although this literature states that it has not been proven if a high or low sway variability is beneficial to balance, we can assume that a pronated foot, with a higher sway variability and more flexibility, provides a greater ability to adapt to somatosensory information. If a subject with a low MLA conducted balance testing on a foam or an air-inflated instability device, we could assume that the subject would be able to maintain postural sway more efficiently than a subject with a high MLA because of the flexibility of the foot and the area of the foot in contact with the device.

The difference in CoP and pressure location during walking for subjects with flat and normal feet was observed (Han et al., 2011). The pressure pathway in the normal feet group was found to move from the lateral heel to the hallux. However, the flat feet group displayed a pathway of pressure that went straight from the heel to the phalanges (Han et al., 2011). The results also displayed that peak plantar pressure of normal feet was significantly higher in the fourth and fifth metatarsals (130.34 ± 68.38 kPa) and heel (247.81 ± 38.48 kPa) than in the fourth and fifth metatarsal (83.65 ± 32.05 kPa) and heel (198.54 ± 30.77 kPa) of the flat feet group (Han et al., 2011).

A similar study by Seguín et al. (2014) observed the differences in plantar pressure for subjects with normal feet and pes cavus (high MLA) when walking (Fernández-Seguín et al., 2014). This study found a significant increase in pressure for the metatarsals (656.12 ± 22.39 kPa) and the forefoot (728.69 ± 24.14 kPa) within the pes cavus group compared to the normal group with a metatarsal pressure of (503.79 ± 9.32 kPa) and a forefoot pressure of (631.36 ± 9.61 kPa). There was a significant decrease in pressure in the hallux for the pes cavus group (56.69 ± 3.05 kPa) compared to the normal group (100.14 ± 3.46 kPa). This study supports the idea that a high arch drives the center of pressure to the anterior and lateral aspects of the foot (Fernández-Seguín et al., 2014). Click or tap here to enter text.

Medical Conditions That Affect Balance

Chronic ankle instability (CAI) is a reoccurring ankle sprain that can significantly impact postural control. An ankle sprain is caused by the overstretching of the ligaments that connect the talocrural and subtalar joints when the foot is in the position of inversion with plantar flexion and internal rotation (Garrick, 1977; Hertel, 2000). Lateral ankle sprains cause elasticity of the lateral ligaments. Because of this, CAI can decrease the ability to maintain postural control (Cornwall & Murrell, 1991; Friden et al., 1989; Gauffin et al., 1988; Goldie et al., 1994; Golomer et al., 1994; Guskiewicz & Perrin, 1996; Leanderson et al., 1993, 1996; Orteza et al., 1992; Perrin et al., 1997; Tropp et al., 1984; Wortmann & Docherty, 2013). The mechanoreceptors of the lateral ligaments are responsible for sensing the overstretching from inversion, but once damaged, an ankle sprain will become more frequent. Peroneal longus and brevis muscles on the lateral side of the lower leg are responsible for the eversion of the foot (Hertel, 2000). After a lateral ankle sprain, the activation time for these muscles is longer (Brunt et al., 1992; Karlsson & Andreasson, 1992; Konradsen & Ravn, 1990; Löfvenberg et al., 1995; Lynch et al., 1996; Richer et al., 2022).

A study supports ankle sprains as a possible injury in subjects with a larger sway velocity while performing a single-leg static balance test (Tropp et al., 1984). Tropp et al. tested the stabilometry of athletes with and without previous ankle injuries while in a single-leg stance on a force platform. The postural sway of the subjects was classified into two categories: normal and pathological. The normal data group consisted of postural sway data points within two standard deviations of the control group. The pathological data group had postural sway data points that exceeded two standard deviations of the reference group. Of the 127 participants in this study, 29 participants had a history of ankle injuries, and 71 participants had no previous ankle injuries. When considering the subjects with pathologic stabilometry results, 12 subjects experienced ankle injuries, and 17 did not previously have ankle injuries. This results in the pathological group having a 42% risk of re-injury. However, the normal stabilometry group had 87 participants with no history of ankle injury and only 11 participants with previous ankle injuries. Subjects with data points within the normal area range only had an 11% ($P<0.001$) chance of re-injury. The study concluded that stabilometry value was the only significant factor determining future ankle injury (Tropp et al., 1984). Subjects with postural sway data points in the pathological category are more at risk for ankle injury regardless of previous medical history.

The modified Rhomberg test or force plates can assess the effects of CAI on postural control. A joint position assessment can be conducted to determine joint laxity, and an electromyograph device can assess muscle activation of the peroneus muscles.

The plantar fascia is a dense collagen band that continues from the Achilles tendon to the metatarsal heads of the foot (Natali et al., 2010). This band of collagen supports the MLA, which serves as a shock absorber from the force exerted by the contact from the ground. Plantar fasciitis is a medical condition in which the plantar fascia obtains microtears from an overload in weight and strain. This produces inflammation in the plantar fascia tissue (Richer et al., 2022). The continuum of strain on collagen fibers will result in a thickening of the tissue. If the plantar fascia is considered stretched due to a high arch, there will be a strain on the Achilles tendon and the gastrocnemius and soleus muscles (Ağırman, 2018; Cheung et al., 2006). Static and dynamic postural control can be affected by the Achilles tendon strain from the plantar fascia (Ağırman, 2018; Gonçalves et al., 2017; Petrofsky et al., 2020).

Unilateral vs. Bilateral Stance

Standing on one foot is defined as a unilateral stance, whereas standing on two feet is a bilateral stance. When addressing stability, there is a natural understanding that a unilateral stance is much more challenging than a bilateral stance. The body's CoM is located in the medial aspect of the trunk. When one side of the base of support is absent, the body must shift the weight to the leg that is supporting the body.

Aside from weight distribution, a unilateral stance can produce decreased stability due to imbalances in strength between the dominant and nondominant legs. The bilateral stance can conceal the muscular deficiencies of the weaker limb. Unilateral balance tests between the dominant and nondominant legs can show a difference in postural sway parameters since the nondominant leg must rely on weaker muscles.

Surface

Instability Device

The use of an instability device can manipulate postural control. It can alter the somatosensory system by providing an unstable surface. Since the somatosensory system is hindered, the CNS cannot make postural adjustments accurately (Boonsinsukh et al., 2020). Many studies implement an instability device to observe the neuromuscular adjustments made by the central nervous system after manipulating the somatosensory system. As previously discussed, instability devices test the vestibular and visual systems when making postural adjustments because the somatosensory system is altered.

The NeuroCom and Airex foam pads are commonly used for rehabilitation clinics and research. The density and elasticity of a foam pad are relevant to the reliability of testing results. If a researcher is curious about postural sway, they may use a foam pad with a high elasticity. The density of the foam pad will vary among subjects due to body weight differences. Research by Boonsinsuhk et al. compared postural sway in different age groups when standing on the ground, the Airex foam pad, and the NeuroCom foam pad accurately (Boonsinsukh et al., 2020). The results revealed that postural sway increased in all age groups when standing on an instability device. However, the difference between the types of foam pads was insignificant (Boonsinsukh et al., 2020). This insignificant difference was likely due to the researchers using two Airex foam pads (0.12 m combined) to imitate the thickness of one NeuroCom foam pad (0.13 m). The density of the combined Airex foam pads was 55 kg(m^{-3}), and the density of the

NeuroCom foam pad was 60 kg(m^{-3}). As one can expect, a foam pad with a high density can decrease postural instability due to its firmness. However, a foam pad with a low density can produce a large amount of postural instability because the foam pad produces more laxity.

Research by Hong et al. observed the specificity and sensitivity between foam pad thickness and visual conditions (Hong et al., 2015). The sensitivity of the data in this experiment is best described as the relationship between the results and the size of the dataset. The specificity of this experiment is described as the quantitative value of effectiveness when used on subjects in a control group. The eyes-opened foam pad test results were 7% (1-26) sensitivity and 100% specificity (92.7-100) for the 12 cm thickness. The 18 cm foam pad thickness had a 15% (5-33) sensitivity percentage and a 100% (92-100) specificity percentage. Results for the eyes-closed foam pad condition were 36% (21-54) sensitivity and 92% (80-98) specificity for 12 cm foam thickness. When the third foam pad was added to reach 18 cm of thickness, the sensitivity percent was 63% (45-78), and the specificity percent was 90% (77-97). This study concluded that the eyes-closed testing condition with three foam pads (18 cm thickness combined) is acceptable for testing postural control systems (Hong et al., 2015).

Another study by Stanek et al. sought to classify the difficulty level in maintaining postural control on four standard instability devices by measuring CoP displacement and average sway velocity (Stanek et al., 2013). An Airex balance pad, BOSU, DynaDisc, and a half-foam roller were the instability devices used in their research. Subjects were instructed to stand on the dominant leg, the other leg flexed at the knee, and hands placed on hips. The baseline displacement measure was 0.01 ± 0.24 cm for ML and -0.03 ± 0.52 cm for AP. The BOSU trainer had an ML displacement of 0.51 ± 0.90 cm and an AP displacement of -0.24 ± 0.91 cm. The DynaDisc displayed an ML displacement of 0.26 ± 0.56 cm and a displacement of -0.07 ± 0.64 cm in the AP direction. The Airex balance pad's displacement was 0.31 ± 0.84 cm for the ML direction and -0.15 ± 1.63 cm for the AP direction. Lastly, the half-foam roller resulted in an ML displacement of 0.11 ± 0.05 cm and a -0.05 ± 0.60 cm AP displacement (Stanek et al., 2013). This research team concluded that the DynaDisc and BOSU trainer were the two most challenging training devices, followed by the Airex foam pad and the half-foam roller when comparing CoP area, average sway velocity, and effect sizes in the ML direction. The results imply that individuals will slightly shift the CoP to the more posterolateral region of the foot. There were no significant differences in the CoP area, average sway velocity, or effect sizes when observing the AP direction (Stanek et al., 2013). This could be expected because when an individual is in a unipedal stance, instability tends to shift more to the ML direction due to the CoM not being directly and equally over the CoP. Since research supports the idea that CoP shifts laterally and posteriorly when standing on an instability device, it would be beneficial to train the activation of the muscles responsible for maintaining postural control when the CoM goes beyond the lateral and posterior CoP boundaries (Stanek et al., 2013).

Training Interventions

Consistently training on an instability device can cause neuromuscular adaptations. Balance training is a common practice for individuals interested in improving postural control. Implementing balance training can increase an individual's ability to adjust to postural disruptions.

The literature presented by Wortmann and Docherty provides several studies that have observed the effects of balance training in subjects with CAI (Wortmann & Docherty, 2013). Three of the studies mentioned resulted in a decrease in single-limb static postural sway after completion of the training protocol. Dynamic postural control was also improved after a six-week training (Wortmann & Docherty, 2013).

Another study by Rossi et al. observed the training period needed to increase postural control in older women (Rossi et al., 2013). The experimental group in this study completed a balance training protocol that consisted of balance exercises that lasted 40 minutes and were completed three days per week for six weeks. The instability devices used were the following: a proprioceptive disk, a rocker, a balance board, a BOSU ball, an inverted BOSU ball, and a mini trampoline (Rossi et al., 2013). Muscular contractions of the tibialis anterior, gastrocnemius, and soleus were measured using electromyography technology (EMG). The results of the training group showed a faster muscle contraction (0-200 ms) in the tibialis anterior (166.85 ± 30.06 ms), gastrocnemius (87.52 ± 16.24 ms), and soleus (84.47 ± 16.13 ms) when needed to maintain postural control and a significant decrease in backward CoP displacements ($p < 0.0001$) compared to pretraining and the control group (Rossi et al., 2013). This experiment also investigated the effect of a detraining period of six weeks. After detraining, the training group had a statistically significant decrease in early EMG activity (0-200 ms) of the tibialis anterior (164.42 ± 26.01 ms), gastrocnemius (79.14 ± 18.03 ms), and soleus (70.14 ± 16.17 ms) compared to the post-training results; however, it was still significantly higher than the control group. There was no significant difference in backward CoP displacement within the training group after detraining compared to post-training results. This study provides data supporting instability device training, which has a relatively chronic effect on CoP displacement and muscle activation (Rossi et al., 2013).

Footwear

When testing postural sway, it is essential to consider the subject's footwear. Although some research states that walking barefoot or with socks is just as risky as wearing insufficient shoes, Menant et al. proposed the idea that the structure of a shoe may affect balance. Some of the projected factors are the following: heel height, heel-collar height, sole hardness, heel and midsole geometry, and slip resistance (Menant et al., 2008). There were six types of shoes used in this experiment to test the concerning factors. The data presented a significant difference in sway when the shoe had an elevated heel (Menant et al., 2008). Elevation of the heel disrupts balance equilibrium by shifting the CoM anteriorly. Lateral instability could also be a consequence due to a smaller tipping angle. To enhance balance in an elderly population, the results from this study suggest a stiff sole or a high-heel-collar type of shoe (Menant et al., 2008). A study by Lord et al. observed similar measures. The experiment tested the postural sway of low and high-heeled shoes, barefoot, and in their shoes (Lord & MBBS, 1996). This study shows similar results to Menant et al.'s; the recommended shoe structure would be a low-heeled or barefoot shoe. In conclusion, the individual's footwear dramatically impacts the ability to control postural sway. Several studies provide evidence that a high-heeled shoe structure creates a large amount of instability.

Shoes can also affect results when using force plates to test postural sway parameters. Asking the participants to perform the tests barefoot or while wearing a sock is common. This allows for a better reading of data points when the foot is in direct contact with the force plates.

CHAPTER III: METHODS

Participants

Thirty healthy (height 167.07 ± 10.7 cm, body mass 67.76 ± 13.45 kg, age 21.67 ± 0.75 years) participants volunteered for this investigation. Inclusionary criteria for this investigation consisted of participants having no neurological disorders and no lower extremity injuries over the previous 12 months. Participants were instructed to refrain from exercise involving the lower extremities for 12 hours before their experimental session. This was confirmed verbally by each participant upon arrival to the session. Lastly, participants completed a physical activity readiness questionnaire (PAR-Q) and provided informed written consent before testing. All procedures were approved by the University of Southern Mississippi institutional review board.

Research Design

A within-subject randomized crossover research design was used to observe differences in sway parameters across conditions. Static balance of the dominant leg was measured on different instability devices during a single session lasting approximately thirty minutes. A control condition required the subject to stand barefoot on the force platform. The experimental conditions used two separate instability devices (The Slack Block, Slackbow LLC, USA, and AirEX Balance-pad, AirEX, Sins, Switzerland).

Protocol

All testing was performed using an in-ground force platform (AMTI, Watertown, MA, USA). Participants were instructed to perform trials barefoot. The participant's dominant leg was established by giving the participant a slight nudge in the back. (Lin et al. 2009) The leg used to step forward was deemed the dominant leg and was used for every trial. Participants were also instructed to keep their arms down by side and away from the body.

Each trial began when the subject raised the nondominant leg off the ground. Three trials were performed in each condition. Trials lasted 10 seconds and were followed by a 10-second rest period. A 5-minute rest period was given between each condition. If participants stepped off the instability device, the nondominant foot touched the ground, or if the hands touched the body, the trial was deemed unsuccessful and then repeated until three successful trials were collected.

Data Analysis

Data was processed through the Balance Clinic Software (AMTI, Watertown, MA, USA). CoP data was used to calculate the following variables. Maximal AP and ML sway displacements (Equation 1). Peak AP and ML sway velocity (Equation 2) and average sway velocity (Equation 3). Lastly, total CoP path length (Equation 4) and 95% ellipse area (Equation 5).

Equation #1

$$x_{max} = max(x_i - x_{avg})$$

Equation #2

$$V_{x\,max} = max\left(\frac{(x_i - x_{i-l})}{dt}\right)$$

Equation #3

$$l_{unit} = \frac{l_{path}}{t}$$

Equation #4

$$L_{path} = \sum_{i=2}^{n} \sqrt{(x_i - x_{i-l})^2 + (y_i - y_{i-l})^2}$$

Equation #5

$$A = \pi * \sqrt{F * (x_{sd}^2 + y_{sd}^2 + D)} * \sqrt{F * (x_{sd}^2 + y_{sd}^2 - D)}$$

Statistical Analysis:

The mean data of the three trials were used in all statistical analyses. A one-way repeated measures analysis of variance was conducted for each variable of interest to compare means across the three conditions. Fisher's least significant differences post hoc comparisons were used in determining where statistical differences were present between conditions. Significance for all tests was a priori set at $p < 0.05$. All statistical analyses were performed using SPSS (v25.0, SPSS., Chicago, IL, USA).

CHAPTER IV: RESULTS

All variables means are presented in Table 1. Statistically significant differences were seen in the sway area between conditions (f(2,42) = 5.28, p = 0.009, η^2 = 0.18) (Figure 6). The control condition (9.64 ± 4.53 cm) was significantly lower than both the foam pad (13.05 cm ± 4.25 cm, p = 0.009) and block (12.33 ± 3.37 cm, p = 0.046). No differences were seen between the foam pad and block conditions (p = 0.395). Similarly, statistically significant differences were seen in CoP path length between conditions (f(2,42) = 5.52, p = 0.007, η^2 = 0.21), with the control (67.51 ± 9.49 cm) being significantly lower than both the foam pad (74.36 cm ± 9.76 cm, p = 0.018) and block (76.38 ± 14.84 cm, p = 0.005) (Figure 6). No differences were seen between the foam pad and block conditions (p = 0.442). Maximal ML CoP displacements were significantly different between conditions (f(2,42) = 6.24, p = 0.004, η^2 = 0.23). Lower displacements were seen in the control (1.39 ± 0.20 cm), which was statistically different from both the foam pad (1.59 ± 0.24 cm, p = 0.002) and block (1.53 ± 0.25 cm, p = 0.03) (Figure 3). No differences were present between the foam pad and block conditions. Average sway velocity displayed s statistically significant difference between conditions (f(2,42) = 5.53, p = 0.007, η^2 = 0.21) (Figure 5). The control condition displayed the lowest average sway velocity (6.75 ± 0.95 cm/s), which was statistically different from both the foam pad (7.44 ± 0.98 cm/s, p = 0.005) and block (7.64 ± 1.48 cm/s, p = 0.018). Maximal AP CoP displacements were not significantly different between conditions (f(2,42) = 1.50, p = 0.23, η^2 = 0.07) (Figure 3). Peak AP sway velocity was not significantly different between conditions (f(2,42) = 1.75, p = 0.186, η^2 = 0.08) (Figure 4). Lastly, peak ML sway velocity was not significantly different between conditions (f(2,42) = 2.16, p = 0.13, η^2 = 0.09) (Figure 4).

Table 1: Sway Parameter Comparisons (mean ± *SD*)

	Control	Foam Pad	Block	*p*	η^2
Peak AP Displacement (cm)	2.05 ± 0.56	2.31 ± 0.62	2.23 ± 0.58	0.23	0.07
Peak ML Displacement (cm)	1.39 ± 0.20	1.59 ± 0.24*	1.53 ± 0.25*	0.004	0.23
Peak AP Sway Velocity (cm/s)	26.66 ± 6.03	28.94 ± 7.45	29.88 ± 6.91	0.19	0.08
Peak ML Sway Velocity (cm/s)	26.35 ± 6.85	28.58 ± 5.81	29.62 ± 5.88	0.13	0.09
Average Sway Velocity (cm/s)	6.75 ± 0.95	7.44 ± 0.98*	7.64 ± 1.48*	0.007	0.21
Path Length (cm)	67.51 ± 9.49	74.36 ± 9.76*	76.38 ± 14.84*	0.007	0.21
Sway Area (cm^2)	9.65 ± 4.53	13.04 ± 4.25*	12.33 ± 3.37*	0.009	0.18

AP = anterior – posterior

ML = medial – lateral

* = significantly different from control condition

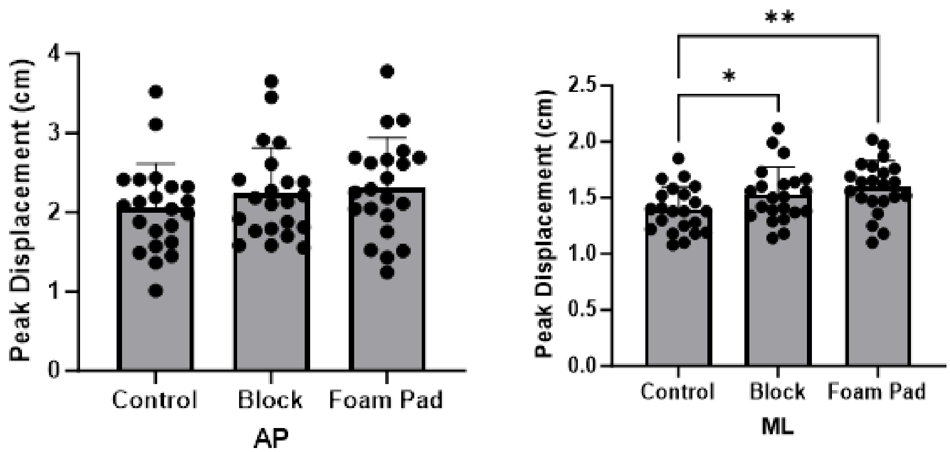

***Figure 3**: Peak CoP Displacement*

* = significantly different from control condition ($p < 0.05$)

** = significantly different from control condition ($p < 0.01$)

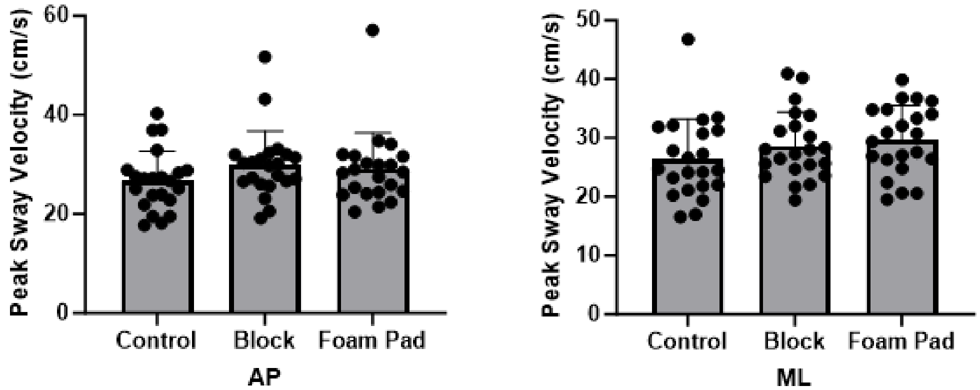

***Figure 4**: Peak Sway Velocity*

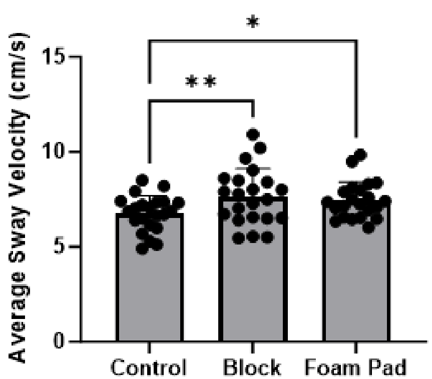

***Figure 5:** Average Sway Velocity*

* = significantly different from control condition ($p < 0.05$)

** = significantly different form control condition ($p < 0.01$)

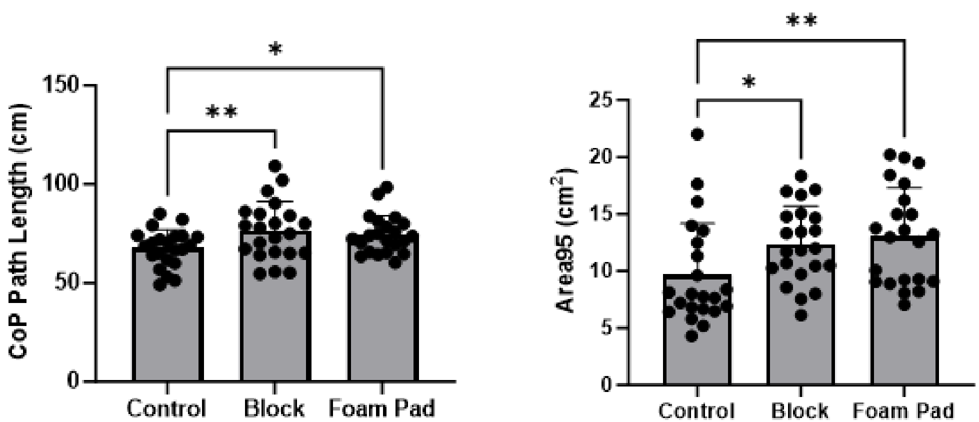

Figure 6: *Cop Path Length and 95th Percentile Ellipse Area*

* = significantly different from control condition ($p < 0.05$)

** = significantly different form control condition ($p < 0.01$)

CHAPTER V: DISCUSSION

. This study aimed to compare sway parameters between instability devices using the block and the foam condition. Another objective of this experiment was to investigate the difference in sway parameters of the instability devices compared to a controlled condition. Our results indicate no significant differences in sway parameters between the foam and the block conditions. However, there was a significant difference in sway parameters between the ground and the instability devices, which supports the second hypothesis.

In theory, increasing the height of the body's CoM will decrease postural control. Previous research has explored the effects of altering the location of the CoM on postural sway. An experiment by Simeonov observed postural sway differences in construction workers using two instability devices (Simeonov, 2001). The heights of the instability devices were three and nine meters high. When using the three-meter instability device, the area of sway and sway velocity increased dramatically compared to the ground. Their results display another significant increase in sway area and velocity when using a height of nine meters (Simeonov, 2001). A study by Ojie et al. used the angle of hanging mass and the projected sway to observe the relationship between sway and the height of the CoM. It was found that the height of the CoM directly correlated to sway (Ojie et al., 2020). They found that as the height increased from 50 cm to 100 cm, the mean of the CoM displacements continually increased. Our results of sway parameters between the foam and the block may not have had a significant outcome because the height difference between the block and the foam was not large enough. The difference in the heights of the instability devices is due to the addition of the wooden piece to the surface of the block. Since our results show no significant differences between the two devices, we can assume that the height difference between the two devices was not enough to raise the CoM to induce an increase in postural sway for the block condition.

Although the block narrowed the BoS, the wooden surface on the block could serve as a mode of stability for the subject. The wooden piece is dense and acts as a firm surface for the human, not allowing the foot to sink into the foam. Literature by Chaikeree et al. observed postural sway due to instability devices with different physical properties. The foam pad with the highest Young's Modulus value significantly increased sway (Chaikeeree et al., 2015). Though foam stiffness was not measured in this investigation, the foam used in the construction of the block is different from that of the foam pad. The addition of the firm surface on top of the block may have negated any differences that foam density might have played.

Our data shows no significant differences in sway parameters between the foam and block conditions (*Table 1*). Lin et al. had similar results when observing the differences between a firm and soft foam pad. They concluded that there were no significant differences between the instability devices, and they could be used interchangeably in a clinical setting (Lin et al., 2015). Boonsinsukh et al. also found similar results since both instability devices produced a similar magnitude of sway in their study (Boonsinsukh et al., 2020). Based on the findings in the current investigation, we support recommendation of using these devices interchangeably in an effort to induce increase postural instability. However, future investigations should examine the impact of training when using similar protocols with these devices.

We also hypothesized that both instability devices would significantly increase sway parameters compared to the control condition. Several studies have supported this claim. Lin et al., as mentioned above, also investigated sway parameters when using instability devices compared to the ground. Their findings were similar to ours since they found a significant increase in postural sway in the foam conditions (Lin et al., 2015). Patel et al. also used the ground as their control condition when analyzing torque variance when standing on unstable devices with different densities (Lin et al., 2015). Since the ground is firm and has no elasticity, it provides more stability than a foam pad. A firm surface allows the body's mass to be distributed to allow the CoM to be directly in line with the CoP in the foot (Lin et al., 2015).

The foam and block conditions had significantly higher values than the control for peak ML displacement ($p = 0.004$), average sway velocity ($p = 0.007$) path length ($p = 0.007$) and sway area ($p = 0.009$). Janura et al. also investigated CoP displacements when standing on the ground and a foam pad. Their results were similar to ours in that the compliant surface produced high CoP sway velocity (Janura et al., 2017). Since the somatosensory system is manipulated using a foam pad, we can assume that postural instability would increase.

This study has limitations. Subjects were not given specific instructions on positioning the nondominant leg during testing. Previous studies have instructed subjects to flex the nondominant knee at 90 degrees. We have considered ecological validity as a limitation of this study. However, this study aimed to imitate practical use in a clinical setting when attempting to induce additional postural instability. Another limitation of this study is the visual field. The visual target on the wall was not at eye level, which could have increased postural sway due to the head's orientation. This experiment was conducted within a healthy population. To qualify for this study, subjects were not to have any lower extremity injuries within the past year.

To further investigate this topic, a training intervention could be implemented to observe the long-term effects of the instability devices. It would be beneficial to examine if training interventions on the instability devices create increased postural control. Since this experiment conducted three trials lasting 10 seconds each, the trials could be extended in future research. Another route this topic could investigate could be the testing of an unhealthy population. Balance training is crucial for elders, neuromuscular rehabilitation, patients with neurodegenerative disorders, chronic ankle instability patients, autistic populations, and amputees. If sway parameters after a longitudinal training program on the foam pad and the block were observed, rehabilitation clinics could benefit by understanding if these devices could be used interchangeably or if one device could be used as a mode of progression when following a rehabilitation protocol.

In conclusion, this study found no statistically significant differences in sway parameters between the block and foam pad. The results from this study support the idea that the block and foam pad can be used interchangeably for acute balance training (<10 seconds) since they do not exert a significant difference in postural instability

APPENDIX A: IRB APPROVAL LETTER

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NOTICE OF INSTITUTIONAL REVIEW BOARD ACTION

The project below has been reviewed by The University of Southern Mississippi Institutional Review Board in accordance with Federal Drug Administration regulations (21 CFR 26, 111), Department of Health and Human Services regulations (45 CFR Part 46), and University Policy to ensure:

- The risks to subjects are minimized and reasonable in relation to the anticipated benefits.
- The selection of subjects is equitable.
- Informed consent is adequate and appropriately documented.
- Where appropriate, the research plan makes adequate provisions for monitoring the data collected to ensure the safety of the subjects.
- Where appropriate, there are adequate provisions to protect the privacy of subjects and to maintain the confidentiality of all data.
- Appropriate additional safeguards have been included to protect vulnerable subjects.
- Any unanticipated, serious, or continuing problems encountered involving risks to subjects must be reported immediately. Problems should be reported to ORI using the Incident form available in InfoEd.
- The period of approval is twelve months. If a project will exceed twelve months, a request should be submitted to ORI using the Renewal form available in InfoEd prior to the expiration date.

PROTOCOL NUMBER:	23-0148
PROJECT TITLE:	Impact of Differing Instability Devices on Postural Sway Parameters
SCHOOL/PROGRAM	Kinesiology
RESEARCHERS:	PI: Paul Donahue Investigators: Donahue, Paul~
IRB COMMITTEE ACTION:	Approved
CATEGORY:	Expedited Category
PERIOD OF APPROVAL:	03-Apr-2023 to 02-Apr-2024

Donald Sacco Jr.

Donald Sacco, Ph.D.
Institutional Review Board Chairperson

REFERENCES

Ağırman, M. (2018). Evaluation of balance and fall risk in patients with plantar fasciitis syndrome. *SiSli Etfal Hastanesi Tip Bulteni / The Medical Bulletin of Sisli Hospital*. https://doi.org/10.14744/SEMB.2018.68736

Alexander, N. B. (1994). Postural Control in Older Adults. *Journal of the American Geriatrics Society*, *42*(1), 93–108. https://doi.org/10.1111/j.1532-5415.1994.tb06081.x

Atik, A. (2014). Flexible flatfootness. *Northern Clinics of Istanbul*, *1*(1), 57–63. https://doi.org/10.14744/nci.2014.29292

Bednarczuk, G., Wiszomirska, I., Rutkowska, I., & Skowroński, W. (2021). Role of vision in static balance in persons with and without visual impairments. *European Journal of Physical and Rehabilitation Medicine*, *57*(4). https://doi.org/10.23736/S1973-9087.21.06425-X

Birinci, T., & Demirbas, S. B. (2017). Relationship between the mobility of medial longitudinal arch and postural control. *Acta Orthopaedica et Traumatologica Turcica*, *51*(3), 233–237. https://doi.org/10.1016/j.aott.2016.11.004

Boonsinsukh, R., Khumnonchai, B., Saengsirisuwan, V., & Chaikeeree, N. (2020). The effect of the type of foam pad used in the modified Clinical Test of Sensory Interaction and Balance (mCTSIB) on the accuracy in identifying older adults with fall history. *Hong Kong Physiotherapy Journal*, *40*(02), 133–143. https://doi.org/10.1142/S1013702520500134

Brunt, R. L., Anderson, J. C., Huntsman, B., & et al. (1992). Postural responses to lateral perturbation in healthy subjects and ankle sprain patients. *Med Sci Sports Exer*, *24*, 171–176.

Chaikeeree, N., Saengsirisuwan, V., Chinsongkram, B., & Boonsinsukh, R. (2015). Interaction of age and foam types used in Clinical Test for Sensory Interaction and Balance (CTSIB). *Gait & Posture*, *41*(1), 313–315. https://doi.org/10.1016/j.gaitpost.2014.09.011

Chang, Y. W., Hung, W., Wu, H., Chiu, Y. C., & Hsu, H. C. (2010). Measurements of foot arch in standing, level walking, vertical jump and sprint start. *International Journal of Sport and Exercise Science*, *2*(2), 31–38.

Cheung, J. T.-M., Zhang, M., & An, K.-N. (2006). Effect of Achilles tendon loading on plantar fascia tension in the standing foot. *Clinical Biomechanics*, *21*(2), 194–203. https://doi.org/10.1016/j.clinbiomech.2005.09.016

Clément, G., Jacquin, T., & Berthoz, A. (n.d.). Habituation of Postural Readjustments Induced by Motion of Visual Scenes. In *Vestibular and Visual Control on Posture and Locomotor Equilibrium* (pp. 99–104). S. Karger AG. https://doi.org/10.1159/000410310

Cornwall, M. W. , & Murrell, P. (1991). Postural sway following inversion sprain of the ankle. *Journal of the American Podiatric Medical Association*, *81*(5), 243–247. https://doi.org/10.7547/87507315-81-5-243

Cote, K. P., Brunet, M. E., Gansneder, B. M., & Shultz, S. J. (2005). Effects of Pronated and Supinated Foot Postures on Static and Dynamic Postural Stability. *Journal of Athletic Training*, *40*, 41–46.

Faquin, B. S., Candido, C. R. C., Mochizuki, L., & Okazaki, V. H. A. (2018). Effect of visual and vestibular information on spatial perception on gait. *Human Movement*, *2018*(2), 39–45. https://doi.org/10.5114/hm.2018.74058

Fernández-Seguín, L. M., Diaz Mancha, J. A., Sánchez Rodríguez, R., Escamilla Martínez, E., Gómez Martín, B., & Ramos Ortega, J. (2014). Comparison of plantar pressures and contact area between normal and cavus foot. *Gait & Posture*, *39*(2), 789–792. https://doi.org/10.1016/j.gaitpost.2013.10.018

Friden, T., Zätterström, R., Lindstrand, A., & Moritz, U. (1989). A stabilometric technique for evaluation of lower limb instabilities. *The American Journal of Sports Medicine*, *17*(1), 118–122. https://doi.org/10.1177/036354658901700120

Fukano, M., & Fukubayashi, T. (2009). Motion characteristics of the medial and lateral longitudinal arch during landing. *European Journal of Applied Physiology*, *105*(3), 387–392. https://doi.org/10.1007/s00421-008-0915-3

Garrick, J. G. (1977). The frequency of injury, mechanism of injury, and epidemiology of ankle sprains*. *The American Journal of Sports Medicine*, *5*(6), 241–242. https://doi.org/10.1177/036354657700500606

Gauffin, H., Tropp, H., & Odenrick, P. (1988). Effect of Ankle Disk Training on Postural Control in Patients with Functional Instability of the Ankle Joint. *International Journal of Sports Medicine*, *09*(02), 141–144. https://doi.org/10.1055/s-2007-1024996

Goldie, P. A., Evans, O. M., & Bach, T. M. (1994). Postural control following inversion injuries of the ankle. *Arch Phys Med Rehab*, *75*, 969–975.

Golomer, E., Dupui, Ph., & Bessou, P. (1994). Spectral frequency analysis of dynamic balance in healthy and injured athletes. *Archives Internationales de Physiologie, de Biochimie et de Biophysique*, *102*(3), 225–229. https://doi.org/10.3109/13813459409007543

Gonçalves, G. A., Kamonseki, D. H., Martinez, B. R., Nascimento, M. A., Lombardi Junior, I., & Yi, L. C. (2017). Static, dynamic balance and functional performance in subjects with and without plantar fasciitis. *Fisioterapia Em Movimento*, *30*(1), 19–27. https://doi.org/10.1590/1980-5918.030.001.ao02

Guskiewicz, K. M., & Perrin, D. H. (1996a). Effect of Orthotics on Postural Sway Following Inversion Ankle Sprain. *Journal of Orthopaedic & Sports Physical Therapy*, *23*(5), 326–331. https://doi.org/10.2519/jospt.1996.23.5.326

Guskiewicz, K. M., & Perrin, D. H. (1996b). Research and clinical applications of assessing balance. *J Sport Rehabil*, *5*, 45–63.

Han, J. T., Koo, H. M., Jung, J. M., Kim, Y. J., & Lee, J. H. (2011). Differences in Plantar Foot Pressure and COP between Flat and Normal Feet During Walking. *Journal of Physical Therapy Science*, *23*(4), 683–685. https://doi.org/10.1589/jpts.23.683

Hertel, J. (n.d.). *Functional Instability Following Lateral Ankle Sprain*.

Hong, S. K., Park, J. H., Kwon, S. Y., Kim, J.-S., & Koo, J.-W. (2015). Clinical efficacy of the Romberg test using a foam pad to identify balance problems: a comparative study with the sensory organization test. *European Archives of Oto-Rhino-Laryngology*, *272*(10), 2741–2747. https://doi.org/10.1007/s00405-014-3273-2

Horak, F. B. (1987). Clinical Measurement of Postural Control in Adults. *Physical Therapy*, *67*(12), 1881–1885. https://doi.org/10.1093/ptj/67.12.1881

Horak, F. B. (1996). Exercise: Regulation and integration of multiple systems. In *Handbook of Physiology* (pp. 255–255).

Horak, F. B. (1997). Clinical assessment of balance disorders. *Gait & Posture*, *6*(1), 76–84. https://doi.org/10.1016/S0966-6362(97)00018-0

Horak, F. B., & Hlavacka, F. (2001). Somatosensory Loss Increases Vestibulospinal Sensitivity. *Journal of Neurophysiology*, *86*(2), 575–585. https://doi.org/10.1152/jn.2001.86.2.575

Horak, F. B., & Nashner, L. M. (1986). Central programming of postural movements: adaptation to altered support-surface configurations. *Journal of Neurophysiology*, *55*(6), 1369–1381. https://doi.org/10.1152/jn.1986.55.6.1369

Jacob, R. G., Redfern, M. S., & Furman, J. M. (1995). Optic flow-induced sway in anxiety disorders associated with space and motion discomfort. *Journal of Anxiety Disorders*, *9*(5), 411–425. https://doi.org/10.1016/0887-6185(95)00021-F

Jacob, R. G., Woody, S. R., Clark, D. B., Lilienfeld, S. O., Hirsch, B. E., Kucera, G. D., Furman, J. M., & Durrant, J. D. (1993). Discomfort with space and motion: A possible marker of vestibular dysfunction assessed by the situational characteristics questionnaire. *Journal of Psychopathology and Behavioral Assessment*, *15*(4), 299–324. https://doi.org/10.1007/BF00965035

Janura, M., Bizovska, L., Svoboda, Z., Cerny, M., & Zemkova, E. (2017). Assessment of postural stability in stable and unstable conditions. *Acta of Bioengineering and Biomechanics*, *19*(4), 89–94.

Karlsson, A., & Persson, T. (1997). The ankle strategy for postural control—a comparison between a model-based and a marker-based method. *Computer Methods and Programs in Biomedicine*, *52*(3), 165–173. https://doi.org/10.1016/S0169-2607(96)01794-4

Karlsson, J., & Andreasson, G. O. (1992). The effect of external ankle support in chronic lateral ankle joint instability. *The American Journal of Sports Medicine*, *20*(3), 257–261. https://doi.org/10.1177/036354659202000304

Konradsen, L., & Ravn, J. B. (1990). Ankle instability caused by prolonged peroneal reaction time. *Acta Orthopaedica Scandinavica*, *61*(5), 388–390. https://doi.org/10.3109/17453679008993546

Leanderson, J., Eriksson, E., Nilsson, C., & Wykman, A. (1996). Proprioception in Classical Ballet Dancers. *The American Journal of Sports Medicine*, *24*(3), 370–374. https://doi.org/10.1177/036354659602400320

Leanderson, J., Wykman, A., & Eriksson, E. (1993). Ankle sprain and postural sway in basketball players. *Knee Surgery, Sports Traumatology, Arthroscopy*, *1*(3–4), 203–205. https://doi.org/10.1007/BF01560207

Lee, P.-Y., Gadareh, K., & Bronstein, A. M. (2014). Forward–backward postural protective stepping responses in young and elderly adults. *Human Movement Science*, *34*, 137–146. https://doi.org/10.1016/j.humov.2013.12.010

Lestienne, F., Soechting, J., & Berthoz, A. (1977). Postural readjustments induced by linear motion of visual scenes. *Experimental Brain Research*, *28–28*(3–4). https://doi.org/10.1007/BF00235717

Lin, C.-C., Roche, J. L., Steed, D. P., Musolino, M. C., Marchetti, G. F., Furman, G. R., Redfern, M. S., & Whitney, S. L. (2015). *Test-retest reliability of postural stability on two different foam pads HHS Public Access* (Vol. 1, Issue 2).

Löfvenberg, R., Kärrholm, J., Sundelin, G., & Ahlgren, O. (1995). Prolonged Reaction Time in Patients with Chronic Lateral Instability of the Ankle. *The American Journal of Sports Medicine*, *23*(4), 414–417. https://doi.org/10.1177/036354659502300407

Lord, S. R., Clark, R. D., & Webster, I. W. (1991). Visual Acuity and Contrast Sensitivity in Relation to Falls in an Elderly Population. *Age and Ageing*, *20*(3), 175–181. https://doi.org/10.1093/ageing/20.3.175

Lord, S. R., & MBBS, G. M. B. (1996). Shoe Characteristics and Balance in Older Women. *Journal of the American Geriatrics Society*, *44*(4), 429–433. https://doi.org/10.1111/j.1532-5415.1996.tb06416.x

Lynch, S. A., Eklund, U., Gottlieb, D., Renstrom, P. A. F. H., & Beynnon, B. (1996). Electromyographic Latency Changes in the Ankle Musculature During Inversion Moments. *The American Journal of Sports Medicine*, *24*(3), 362–369. https://doi.org/10.1177/036354659602400319

Mancini, M., & Horak, F. B. (2010). The relevance of clinical balance assessment tools to differentiate balance deficits. *European Journal of Physical and Rehabilitation Medicine*, *46*(2), 239–248.

Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., & Lord, S. R. (2008). Effects of Footwear Features on Balance and Stepping in Older People. *Gerontology*, *54*(1), 18–23. https://doi.org/10.1159/000115850

Mergner, T., & Rosemeier, T. (1998). Interaction of vestibular, somatosensory and visual signals for postural control and motion perception under terrestrial and microgravity conditions—a conceptual model. *Brain Research Reviews*, *28*(1–2), 118–135. https://doi.org/10.1016/S0165-0173(98)00032-0

Mille, M.-L., Johnson, M. E., Martinez, K. M., & Rogers, M. W. (2005). Age-dependent differences in lateral balance recovery through protective stepping. *Clinical Biomechanics*, *20*(6), 607–616. https://doi.org/10.1016/j.clinbiomech.2005.03.004

Mølgaard, C., Lundbye-Christensen, S., & Simonsen, O. (2010). High prevalence of foot problems in the Danish population: A survey of causes and associations. *The Foot*, *20*(1), 7–11. https://doi.org/10.1016/j.foot.2010.03.002

Natali, A. N., Pavan, P. G., & Stecco, C. (2010). A constitutive model for the mechanical characterization of the plantar fascia. *Connective Tissue Research*, *51*(5), 337–346. https://doi.org/10.3109/03008200903389127

Nawoczenski, D. A., Cook, T. M., & Saltzman, C. L. (1995). The effect of foot orthotics on three-dimensional kinematics of the leg and rearfoot during running. *The Journal of Orthopaedic and Sports Physical Therapy*, *21*(6), 317–327.

Oatis, C. A. (2009). *Kinesiology the mechanics and pathomechanics of human movement*. Wolters Kluwer.

Ojie, O. O. D., Saatchi, R., & Saatchi, M. (2020). Demonstration of the Effect of Centre of Mass Height on Postural Sway Using Accelerometry for Balance Analysis. *Technologies*, *8*(2), 20. https://doi.org/10.3390/technologies8020020

Orteza, L. C. , Vogelbach, W. D. , & Denegar, C. R. (1992). The effect of molded and unmolded orthotics on balance and pain while jogging following inversion ankle sprain. *Journal of Athletic Training*, *27*(1), 80–84.

Perrin, P., Béné, M., Perrin, C., & Durupt, D. (1997). Ankle Trauma Significantly Impairs Posture Control -A Study in Basketball Players and Controls. *International Journal of Sports Medicine*, *18*(05), 387–392. https://doi.org/10.1055/s-2007-972651

Peterka, RobertJ., & Benolken, MarthaS. (1995). Role of somatosensory and vestibular cues in attenuating visually induced human postural sway. *Experimental Brain Research*, *105*(1). https://doi.org/10.1007/BF00242186

Petrofsky, J., Donatelli, R., Laymon, M., & Lee, H. (2020). Greater Postural Sway and Tremor during Balance Tasks in Patients with Plantar Fasciitis Compared to Age-Matched Controls. *Healthcare*, *8*(3), 219. https://doi.org/10.3390/healthcare8030219

Pollock, A. S., Durward, B. R., Rowe, P. J., & Paul, J. P. (2000). What is balance? *Clinical Rehabilitation*, *14*(4), 402–406. https://doi.org/10.1191/0269215500cr342oa

Redfern, M. S., & Furman, J. M. (1994). Postural sway of patients with vestibular disorders during optic flow. *Journal of Vestibular Research : Equilibrium & Orientation*, *4*(3), 221–230.

Redfern, M. S., Yardley, L., & Bronstein, A. M. (2001). Visual influences on balance. *Journal of Anxiety Disorders*, *15*(1–2), 81–94. https://doi.org/10.1016/S0887-6185(00)00043-8

Richer, L., Fortin, E., Gagnon, G., Ngomo, S., Fernandes, K. B. P., Cortis, C., Sobczak, S., & da Silva, R. A. (2022). Impact of plantar fasciitis on postural control and walking in young middle-aged adults. *The Foot*, *53*, 101951. https://doi.org/10.1016/j.foot.2022.101951

Rossi, L. P., Pereira, R., Brandalize, M., & Gomes, A. R. S. (2013). The Effects of a Perturbation-Based Balance Training on the Reactive Neuromuscular Control in Community- -Dwelling Older Women: A Randomized Controlled Trial. *Human Movement*, *14*(3). https://doi.org/10.2478/humo-2013-0029

Runge, C. F., Shupert, C. L., Horak, F. B., & Zajac, F. E. (1999). Ankle and hip postural strategies defined by joint torques. *Gait & Posture*, *10*(2), 161–170. https://doi.org/10.1016/S0966-6362(99)00032-6

Simeonov, P. (2001). Height, surface firmness, and visual reference effects on balance control. *Injury Prevention*, *7*(90001), 50i–553. https://doi.org/10.1136/ip.7.suppl_1.i50

Stanek, J. M., Meyer, J., & Lynall, R. (2013). Single-Limb-Balance Difficulty on 4 Commonly Used Rehabilitation Devices. *Journal of Sport Rehabilitation*, *22*(4), 288–295. https://doi.org/10.1123/jsr.22.4.288

Tanaka, H., & Uetake, T. (2005). Characteristics of postural sway in older adults standing on a soft surface. *Journal of Human Ergology*, *34*(1–2), 35–40.

Tropp, H., Eckstrand, J., & Gillquist, J. (1984). Stabilometry in functional instability of the ankle and its value in predicting injury. *Med Sci Sports Exer*, *16*, 64–66.

Vuillerme, N., Danion, F., Marin, L., Boyadjian, A., Prieur, J. M., Weise, I., & Nougier, V. (2001). The effect of expertise in gymnastics on postural control. *Neuroscience Letters*, *303*(2), 83–86. https://doi.org/10.1016/S0304-3940(01)01722-0

Vuillerme, N., & Nougier, V. (2004). Attentional demand for regulating postural sway: the effect of expertise in gymnastics. *Brain Research Bulletin*, *63*(2), 161–165. https://doi.org/10.1016/j.brainresbull.2004.02.006

Williams, D. S. B., Tierney, R. N., & Butler, R. J. (2014). Increased Medial Longitudinal Arch Mobility, Lower Extremity Kinematics, and Ground Reaction Forces in High-Arched Runners. *Journal of Athletic Training*, *49*(3), 290–296. https://doi.org/10.4085/1062-6050-49.3.05

Williams, D. S., McClay, I. S., Hamill, J., & Buchanan, T. S. (2001). Lower extremity kinematic and kinetic differences in runners with high and low arches. *Journal of Applied Biomechanics*, *17*(2), 153–163.

Winter, D. (1995). Human balance and posture control during standing and walking. *Gait & Posture*, *3*(4), 193–214. https://doi.org/10.1016/0966-6362(96)82849-9

Wortmann, M. A., & Docherty, C. L. (2013). Effect of Balance Training on Postural Stability in Subjects With Chronic Ankle Instability. *Journal of Sport Rehabilitation*, *22*(2), 143–149. https://doi.org/10.1123/jsr.22.2.143

Wu, G., & Chiang, J.-H. (1997). The significance of somatosensory stimulations to the human foot in the control of postural reflexes. *Experimental Brain Research*, *114*(1), 163–169. https://doi.org/10.1007/PL00005616